CONSTANT AND VARIABLE STIFFNESS FOR HUMAN LEG JOINTS IN NORMAL WALKING

A. Selk Ghafari, A. Meghdari, G.R. Vossoughi
Center of Excellence in Design, Robotics and Automation (CEDRA)
School of Mechanical Engineering, Sharif University of Technology
P.O. Box 11365-9567, Tehran, Iran
E-mail: a_selkghafari@mech.sharif.edu

Abstract- **The present study deals with the constant and variable stiffness of the human leg joints, during the normal walking. A seven-link segment model was employed to represent the anterior-posterior motion study of the human body dynamics. The segments were interconnected by damped torsional springs for modeling the joint muscle-tendon complex. Lagrangian mechanics approach employed to representing equations of motion. The stiffness and damping for ankle, knee and hip joints can be estimated using numerical solution and optimization procedure. Damping was found negligible. There was closer agreement between constant joint stiffness given in the literature and the investigated values for stiffness coefficients from proposed model. The model presented may be provide an effective tool for future designing an active exoskeleton system to assist a paraplegic subject to walk by controlling the additional hip, knee and ankle joint stiffness and damping.**

I. INTRODUCTION

Researches into human gait have a wide range of application in medicine, ergonomics, sport science and technology. Most often methods of multibody dynamics are used when investigation is focused on mechanical aspects of the gait [1-3]. Some multibody models for gait simulation represent bipeds of a very simple, not anthropometric kinematic structure [4], other models are limited to two dimensions [5-8]. Walking is a special kind of the multibody system motion with the occurrence of impacts, friction and slipping. These entire phenomenon observed during the foot-ground contact have to be introduced into the model to make it realistic. In some multibody models of the biped the foot-ground contact is modeled by the additional kinematic joints [5].

With the development of the biomechanical models of human body motion, it has become possible to simulate human performance in order to gain insight into intermuscular coordination and to elucidate control strategies of the musculoskeletal systems. A common method to deal with this type of problems is to lump together elements of the human body, e.g., muscles, tendons, ligaments, bones and joints so that the overall musculoskeletal system is represented as a damped elastic mechanism.

Several methods describing the landing phase of running, hopping, or jumping can be found in the literature [9-13]. These models are usually characterized by the presence of elastic springs and viscous dampers, with constant properties and provide a reasonable prediction of the maximal vertical foot-ground reaction force. Depending on the range of joint flexion and on the frequency of motion, a considerable amount of elastic energy can be stored and reused. It has been shown that the dissipated energy in muscles increase when the amplitudes of joint movements are increased [14]. During the ground contact period of running, hopping and trotting, the leg was modeled as a single linear spring or in terms of the leg joints as constant torsional springs for the ankle, knee and hip joints, with no damping. The stiffness of these springs are considered the average value as the ratio between overall force or moment change to overall vertical displacement or angle change for the leg and joint stiffness, respectively [12, 15-17]. Physiologically, the conception of constant mechanical stiffness may not be applicable. For instance, muscular activation, behaved to be directly related to joint stiffness, varies during the stance phase in gait. Human joints and muscle-tendon complex have the ability to store, absorb and release the energy; therefore its stiffness generally depends on the activation level of the muscle. Thus it can be expected that the joint stiffness and damping is nonlinear in nature and that a model accounting for these facts may improve the human dynamical modeling and performance evaluation.

Recently, there has been an increasing interest in the ability to control paraplegic stance. If individuals with paraplegia were able to stand comfortably for a prolonged period of time, it would have many therapeutic, psychological and practical advantages. Much of the researches regarding the control of paraplegic standing have been focused on the use of artificial stimulation to active paralyzed muscles and to partially restore motor function [18]. Activation of muscles either around the ankle, or the hip, or both assists the paralyzed individual in keeping the body's center of gravity within its base of support when stability is perturbed [19]. Paraplegic standing supported by FES-controlled ankle stiffness is investigated in [20-22], in this study the knee and hip of the paraplegic subject were held in extended positions by a mechanical apparatus, and the closed loop control system maintain the appropriate level of stiffness around the ankle joint. In the other study, evaluation of the additional hip joint stiffness on crutch supported paraplegic stance is investigated [23-24]. In a study illustrated in [25], the role of combined hip and ankle strategy during standing was implemented in a model as an optimal feedback postural control system.

The main goal of the present work was to study the stiffness and damping characteristics of the human joints in normal walking. The present study was thus pretended to provide an insight to realistic joint characteristic using the experimental gait data. It may be provide an effective tool for future designing an active exoskeleton system to assist a paraplegic subject to walk by controlling the additional hip, knee and ankle joint stiffness and damping.

II. MATHEMATICAL MODEL OF THE HUMAN BODY IN WALKING

A seven-link segment model is used to represent the human body gait dynamics as depicted in Fig. 1(a). The seven segments represent the left and right thighs, the shanks and

the feet, in addition, the head, arms and trunk (HAT) are illustrated by one segment. The mass, position of center of mass, moment of inertia, and length of each of the segments are scaled to the dimensions of a human (body mass 80 kg and 1.80 m height) using anthropometric data given in [26]. The segments are connected by torsional spring and damper for modeling the joint muscle-tendon complex. The model is planar and accounts only for anterior-posterior motion study of the body. For convenience, it is assumed that, initially, the right foot is forward with the right heel striking the floor. The joint moments are denoted by subscripted M's, with the first subscript indicating the right or left side of the body, and the second subscript indicating the ankle, knee and hip joints. Each of the feet is represented by a triangle of appropriate shape, as depicted in Fig. 1(b).

The different phases of walking require two basically different mathematical representations. The different phases of walking are [3, 27]:

Phase 1: From right-heel-strike (RHS) to left-toe-off (LTO). Here the right heel and left toe are pivoting on the floor.

Phase 2: From LTO to right-foot-flat (RFF). Only the right heel is pivoting on the floor.

Phase 3: During RFF. Here, the right foot is not moving, and the remaining six segments are pivoting on the right ankle.

Phase 4: From RHO to left-heel-strike (LHS). Only the right toe is pivoting on the floor.

During Phase 1, where six variables describing the two lower limbs are not independent, therefore as suggested in [27], the spring-damper combination in the leg joints plays vital role in the dynamic modeling, but during the swing phase, Phase 2, 3 and 4, the role of these elements were insignificant. In the present study the equations of motion for Phase 1 were considered and the detailed equations of motion for the latter cases may be found in [27].

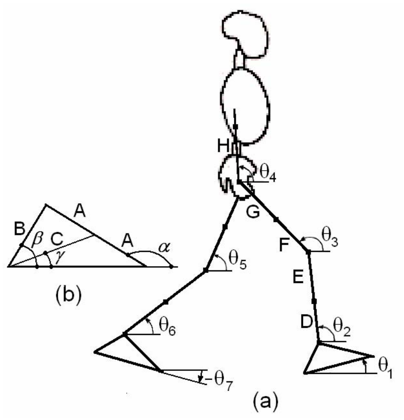

Fig. 1: Link segment model (a) limb angles and limb length variables (b) details of the foot variables

Lagrangian mechanics provides the most direct approach for obtaining equations of motion in this case. The Lagrangian mechanics approach requires the formulation of the Lagrangian given by:

$$L = T - V \qquad (1)$$

Lagrangian written in terms of the independent variables, θ_i, therefore the resulting equation are given as follows:

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{\theta}_i}\right) - \frac{\partial L}{\partial \theta_i} + \frac{F}{\partial \dot{\theta}_i} = Q_{nc,i}, \qquad i = 1, 2, ..., 7 \qquad (2)$$

The detailed formulation of the Lagrangian in Equation (1) and (2) are omitted because the intermediated expressions are very long, instead only the final equation of motion will be given.

The equations of motion for double support case, RHS to LTO, arranged such that only the second derivatives are kept in the left hand side, has the following form:

$$\begin{aligned} &[a_{ij}]\ddot{\theta}_i = b_i \qquad i = 1, 2, ..., 7;\ j = 1, 2, ..., 7 \\ &a_{ij} = c_{ij}\cos(\theta_i - \theta_j) \end{aligned} \qquad (3)$$

$$b_i = -\sum_{j=1}^{7} c_{ij}\dot{\theta}_j^2 \sin(\theta_i - \theta_j) + R_i \qquad (4)$$

Since the c_{ij} are symmetrical, this makes the a_{ij} symmetrical also. On the right hand side of the Equation (3), the b_i are contain moments caused by the spring-damper combination, moments caused by gravity, the applied joint moments and the moments caused by centrifugal forces acting on the limbs. The expressions for c_{ij} and b_i are given in Table I. Also, the c_{ij} not given in the Table I are zero and all are symmetrical.

TABLE I
c_{ij} and R_i COEFFICIENTS IN EQUATIONS (3) AND (4)

$$c_{11} = m_f C^2 + I_f + (m_s + m_t)B^2$$
$$c_{12} = m_s DB + m_t(D + E)B$$
$$c_{13} = m_t FB$$
$$c_{22} = I_s + m_s D^2 + m_t(D + E)^2$$
$$c_{23} = m_t F(D + E)$$
$$c_{33} = I_t + m_t F^2$$
$$c_{44} = I_h + m_h H^2$$
$$c_{45} = m_h H(F + G)$$
$$c_{46} = m_h H(D + E)$$
$$c_{47} = 2m_h HA$$
$$c_{55} = I_t + m_t F^2 + m_h(F + G)^2$$
$$c_{56} = m_t F(D + E) + m_h(D + E)(F + G)$$
$$c_{57} = 2m_t FA + 2m_h A(F + G)$$
$$c_{66} = I_s + m_s D^2 + (m_h + m_t)(D + E)^2$$
$$c_{67} = 2(m_h + m_t)(D + E)A + 2m_s AD$$
$$c_{77} = I_f + 4(0.25m_f + m_h + m_t + m_s)A^2$$
$$R_1 = -m_f gC\cos(\theta_1 - \beta + \gamma) - (m_s + m_t)gB\cos\theta_1 + M_{ra} - C_a\dot{\theta}_1 - K_a\theta_1$$
$$R_2 = -c_{12}g\cos\theta_2 + M_{rk} - M_{ra} - C_k(\dot{\theta}_2 - \dot{\theta}_1) - K_k(\theta_2 - \theta_1)$$
$$R_3 = -c_{13}g\cos\theta_3 + M_{rh} - M_{rk} - C_h(\dot{\theta}_3 - \dot{\theta}_2) - K_h(\theta_3 - \theta_2)$$
$$R_4 = -m_h Hg\cos\theta_4 - M_{rh} - M_{lh} - C_B\dot{\theta}_4 - K_B\theta_4$$
$$R_5 = -[m_h(F + G) + m_t F]g\cos\theta_5 + M_{lh} - M_{lk} - C_h(\dot{\theta}_5 - \dot{\theta}_6) - K_h(\theta_5 - \theta_6)$$
$$R_6 = -[(m_h + m_t)(D + E) + m_s D]g\cos\theta_6 + M_{lk} - M_{la} - C_k(\dot{\theta}_6 - \dot{\theta}_7) - K_k(\theta_6 - \theta_7)$$
$$R_7 = -[2A(m_h + m_t + m_s) + Am_f]g\cos\theta_7 + M_{la} - C_a\dot{\theta}_7 - K_a\theta_7$$

III. Estimation of the Joint Stiffness and Damping Using Numerical Solution

In normal gait there is a natural symmetry between the left and right sides, this is graphically illustrated in Fig. 2. Therefore RHS to LTO takes place during 50% to 62% of gait cycle. Dynamic simulation of the proposed model

requires the knowledge of human kinetics and kinematics data. Joint angles and torques for a typical walking cycle were obtained in the form of independently collected Clinical Gait Analysis (CGA) data. CGA angle data is typically collected via human video motion capture. CGA torque data is calculated by estimating limb masses and inertias and applying dynamic equations to the motion data. Given the variations in individual gait and measuring methods, two independent sources of CGA data [31-32] were utilized for the analysis and estimation of the leg joint stiffness and damping. The anthropometric data required for simulation of the Equation (3) are specified in Table II [27]. In this case spring-damper coefficients of the leg joints are unknown in the proposed model. Numerical solution of the equations of motion in RHS to LTO phase result stiffness-gait cycle percent curves for ankle, knee and hip joints. Simulation results demonstrate angle dependent stiffness properties for joints during normal walking. Ankle, knee and hip stiffness profiles are depicted in Figs. 3 to 5 respectively. Damping coefficients of the joints were found negligible. Average values of investigated leg joints stiffness using numerical solution method are given in Table III and were compared with average values given in the literature [28-30].

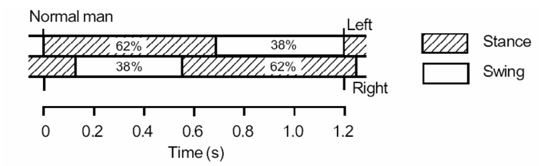

Fig. 2: The time spend on each limb during the normal gait cycle [3]

TABLE II
ANTROPOMETRIC DATA FOR HUMAN BODY SEGMENT

A	Ankle to CG distance	0.116
B	Heel to ankle distance	0.140
C	Heel to CG distance	0.189
D	Ankle to shank CG distance	0.247
E	Shank CG to knee distance	0.188
F	Knee to thigh CG distance	0.227
G	Thigh CG to hip distance	0.173
H	Hip to CG of head, arms and trunk (HAT) distance	0.325
α	Angle between sole and ankle measured at the toe	2.621
β	Angle between sole and ankle measured at the heel	0.967
γ	Angle between sole and CG measured at the heel	0.310
m_f	Foot mass	1.159
I_f	Foot moment of inertia about CG	0.010
m_s	Shank mass	3.717
I_s	Shank moment of inertia about CG	0.064
m_t	Thigh mass	7.994
I_t	Thigh moment of inertia about CG	0.133
m_h	Hat mass	54.2
I_h	HAT moment of inertia about CG	3.591
W_m	Total body mass	79.9
g	Gravitational constant	9.807

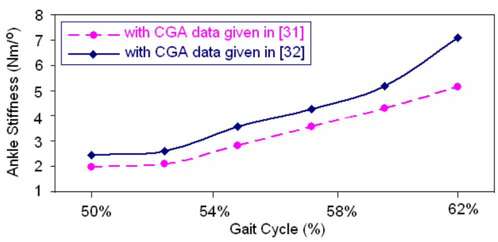

Fig. 3: Ankle stiffness-Gait cycle percent during normal walking

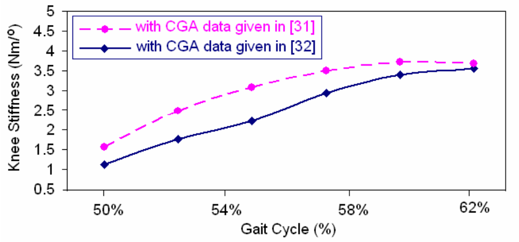

Fig. 4: Knee stiffness-Gait cycle percent during normal walking

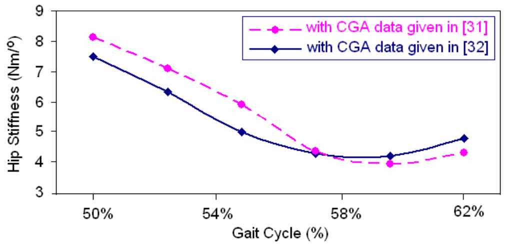

Fig. 5: Hip stiffness-Gait cycle percent during normal walking

IV. ESTIMATION OF THE JOINT STIFFNESS AND DAMPING USING OPTIMIZATION PROCEDURE

The main goal in this section is estimation of the joints stiffness and damping using optimization procedure. Similar to previous section two independent sources of CGA data [31-32] were utilized for the analysis.

Elastic and damping torques are obtained by:

$$\begin{cases} \int_{M_{si0}}^{M_{sij}} dM_{si} = \int_{\theta_{i0}}^{\theta_{ij}} K_i d\theta_i \\ \int_{M_{di0}}^{M_{dij}} dM_{di} = \int_{\omega_{i0}}^{\omega_{ij}} C_i d\omega_i \end{cases} \qquad j = 1,\ldots,n \ , \ i = 1,2,3 \qquad (5)$$

here index i indicated the joint and n is the length of the joint moment vector. The stiffness and damping coefficients in Equation (3) can be solved by parameter estimation using optimization procedure. Minimization of the following objective function J_i was thus performed for each of the joints (ankle, knee and hip).

$$J_i = \sum_{j=1}^{n} (\tau_{ji} - M_{ji})^2 \qquad i = 1,2,3 \qquad (6)$$

here τ_i is CGA torque data. Four inequality constraints were applied as follows. The first two assign positive values for stiffness and damping, thus:

$$\begin{cases} K_i \geq 0 \\ C_i \geq 0 \end{cases} \qquad i = 1,2,3 \qquad (7)$$

The two next constraints were related to the potential elastic energy E_{si} of the spring, and dissipating energy E_{di} of the damper, thus we have:

$$\begin{cases} \int_{E_{si0}}^{E_{sij}} dE_{si} = \int_{\theta_{i0}}^{\theta_{ij}} M_{si} d\theta_i \geq 0 \\ \int_{E_{di0}}^{E_{dij}} dE_{di} = \int_{\omega_{i0}}^{\omega_{ij}} M_{di} d\omega_i \leq 0 \end{cases} \qquad j = 1,\ldots,n \ , \ i = 1,2,3 \qquad (8)$$

Parameter identification was performed by using Quadratic Programming (QP) method [33] to estimate the stiffness and damping of the ankle, knee and hip joints. Damping coefficients of the joints were found negligible using this

method too. Investigated values for joint stiffness coefficients using optimization procedure are given in the Table III. As illustrated in this table, there was closer agreement between constant joint stiffness given in the literature and the investigated values for stiffness and damping coefficients from proposed model using two different methods.

TABLE III
COMPARISON OF THE LEG JOINT STIFFNESS VALUES IN THIS STUDY AND VALUES GIVEN IN THE LITRATURES

	Ankle (Nm/°)	Knee (Nm/°)	Hip (Nm/°)
Stiffness given in [28]	4.69	2.67	2.23
Stiffness given in [29]	4.23	5.65	3.96
Stiffness given in [30]	6.13	5.03	6.23
Optimization results with CGA data [31]	3.11	2.12	3.60
Optimization results with CGA data [32]	4.87	3.39	2.54
Numerical solution with CGA data [31]	3.32	3.01	5.62
Numerical solution with CGA data [32]	4.2	2.50	5.34

IV. CONCLUSIONS AND DISCUSSIONS

The basic outcome of this study is estimation of leg joints stiffness and damping during normal walking. The value of the joints' damping coefficients was very small and negligible. There was closer agreement between constant joint stiffness given in the literature and the investigated values for stiffness and damping coefficients from proposed model using two different methods. FES-compensated ankle, knee and hip stiffness has been utilized to stand up paraplegic patients. By investigating the joints' spring-damper coefficients during normal walking, the presented model provides an effective tool for future designing of active artificial legs, orthosis or development of new walking control strategies with exoskeleton systems for paraplegic patients.

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